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(54) Computed tomography with selectable image resolution

(57) A computed tomography system x-ray detector has a central group of half-width detector elements and groups of full-width elements on each side of the central group. To obtain x-ray attenuation data for whole body layers, the half-width elements are switched effectively into paralleled pairs so all elements act like full-width elements and an image of normal resolution is obtained. For narrower head layers, the elements in the central group are used as half-width elements so resolution which is twice as great as normal is obtained. The central group is also used in the half-width mode and the outside groups are used in the full-width mode to obtain a high resolution image of a body zone within a full body layer. In one embodiment data signals from the detector are switched by electronic multiplexing and in another embodiment a processor chooses the signals for the various kinds of images that are to be reconstructed.

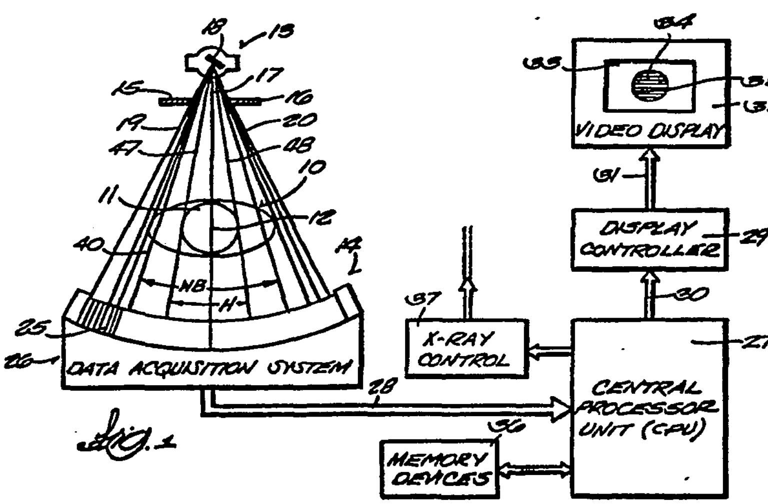

Fig. 1

The drawings originally filed were informal and the print here reproduced is taken from a later filed formal copy.

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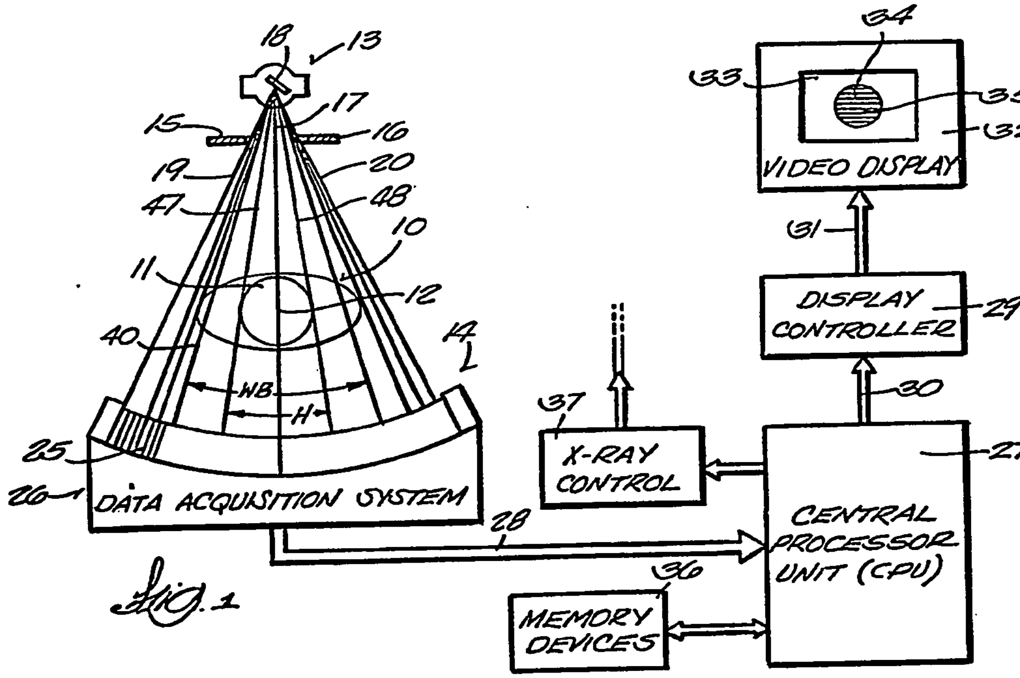

Fig. 1

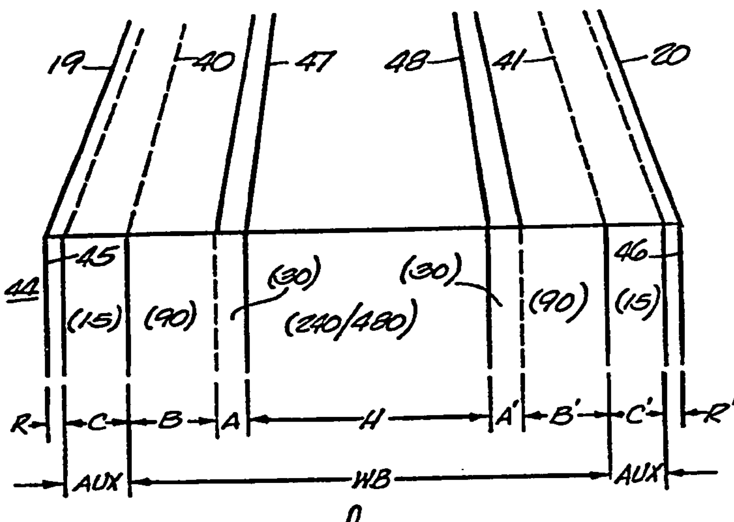

Fig. 2

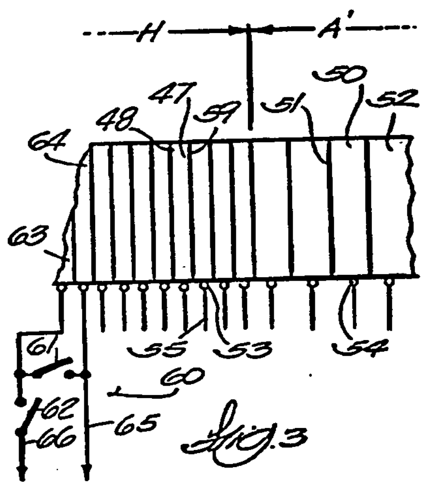

Fig. 3

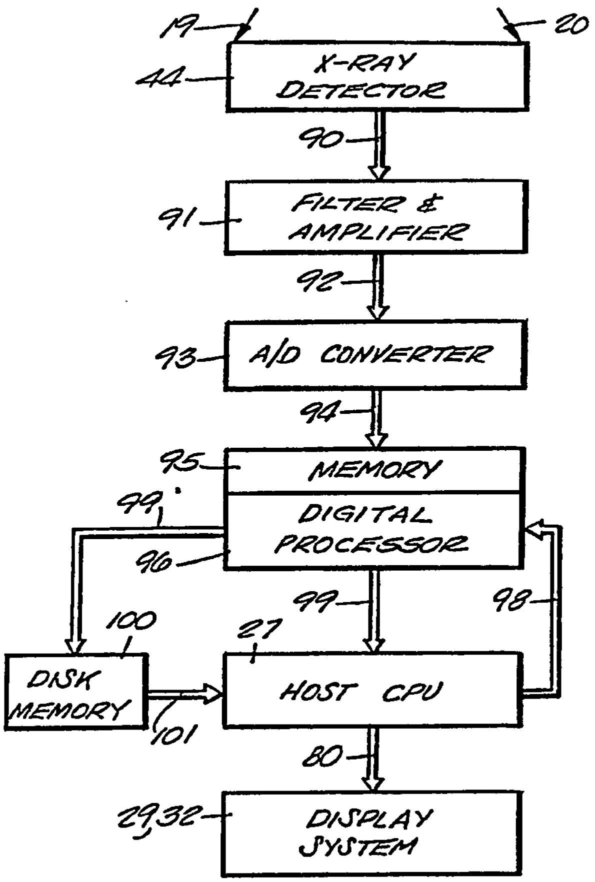

Fig.5

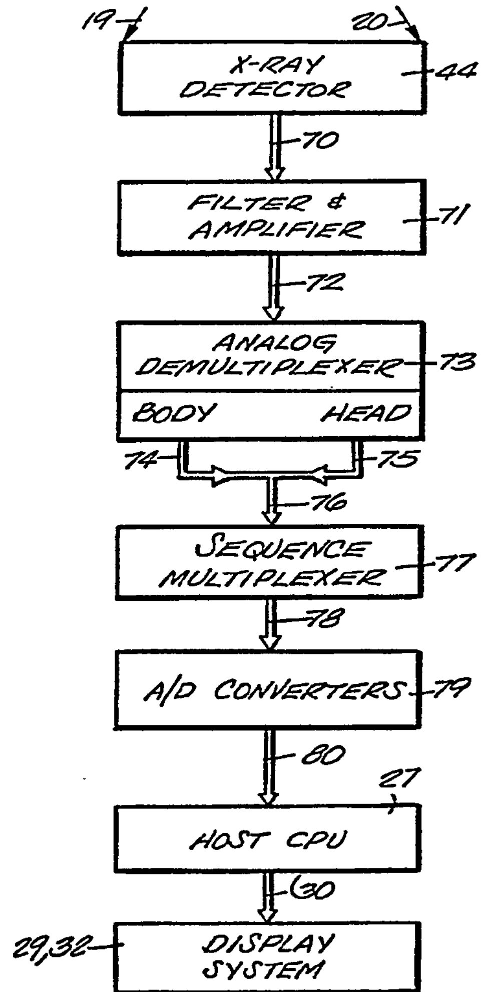

Fig.4

Fig.6

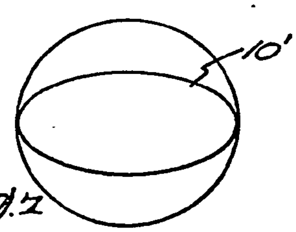

Fig.7

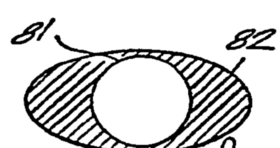

Fig.8

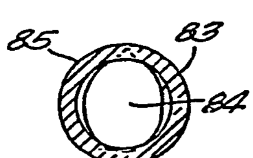

Fig.9

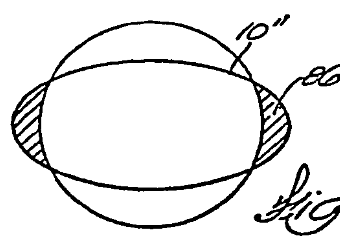

Fig.10

SPECIFICATION

Improvem nts in computed tomography image resolution

This invention is concerned with improving the resolution and images of body layers obtained with x-ray computed tomography systems (hereinafter sometimes called CT and CT systems for the sake of brevity).

In the computed axial tomography process, a spatial distribution of x-ray photon intensities emerging transversely from a layer in a body under examination are translated into analog electric signals which are processed in a manner that enables displaying an axial view of the layer.

In a typical computed tomography scanning apparatus, a body that is to be examined is disposed along a longitudinal axis. An x-ray tube and a multi-element x-ray detector are mounted for orbiting jointly about the axis. The x-radiation which is projected from the x-ray tube through the body to the x-ray detector element array is usually collimated into a fan-shaped beam which is thin in the direction to which the longitudinal axis is perpendicular. The fan-shaped beam, after it emerges from the body, typically diverges sufficiently to spread out over the entire length of the detector element array. In some designs, x-ray attenuation data for enabling reconstruction of a body layer is obtained by pulsing the x-ray tube on and off as it rotates jointly with the detector element array so that x-ray views can be taken at a large number of rotational angles. Each time the x-ray tube is pulsed on, the x-ray photons along individual ray paths which comprise the x-ray beam are detected simultaneously and an analog signal is produced by each detector element which corresponds with the x-ray attenuation along a ray path at the instant of detection. Additional sets of signals are obtained for a sequence of angular positions of the orbiting detector and x-ray source. The discrete analog signals are converted to digital signals and processed in a computer which is controlled by a suitable algorithm to produce signals representative of the x-ray absorption or attenuation of each small volume element in the body through which the x-ray beam passes.

Basically, resolution or revelation of detail in the reconstructed visible image depends on the number of small volumes or picture elements (pixels) of which the visible image is comprised and this, in turn, usually depends on the numb r of discrete elements in the det ctor array. In most CT scanner systems, the divergence of the fan-shaped x-ray beam is such that when a view of a layer in the wide part of a human body is being taken, the differentially attenuated rays which emerge from the body spread across all of the adjacent detector elements comprising the detector array and most of th se elements will be involv d in producing analog signals for enabling reconstruction of an image of the body layer. On some occasions an image of a small infant body or a narrow part of a body such as an arm, a leg, the torso or the head is desired. Considering the head as an example, since it is narrower than the whole body size expected, the projected image of the head will fall on fewer of the detector elements than a projection of the whole body. Since the detector elements which are used for producing x-ray attenuation data for image reconstruction have been customarily all of the same size, resolution of the reconstructed image has been essentially the same for the whole body and the narrower head parts. It is evident that for narrow head or body layers, many of the detector elements and data processing channels in the data acquisition system are inactive and contribute nothing to maximizing resolution of the head or any narrow body part image.

One proposal for utilizing the full capacity of the detector, data acquisition system channels, the computer and the display system for layers of less than full body width such as head layers is to move the x-ray source and its focal spot closer to the head and to move the detector element array jointly with the tube a corresponding distance away from the head which results in the image or shadow of the head extending over the entire length of the detector element array. In this way, all of the detector elements and x-ray attenuation data acquisition channels can be used. This is the approach taken in U.S. Patent No. 4,115,696 which is assigned to the assignee of this application. One of the disadvantages of this proposal is that the x-ray scanner becomes more complicated and costly since it is necessary to provide additional mechanism for shifting the x-ray tube and detector.

Another proposal which has been put into practice is to obtain higher resolution for heads than wider bodies by moving a higher resolution detector array into the beam path when views of a head or other narrow body part are being taken. The substitute detector for heads has detector elements which are about one-half as wide as those which are used in the whole body detector. This results in the image of the head being projected over substantially all of the length of the substitute detector element array and permits all of the data acquisition channels to be used in which case resolution of head layer images become about equal to resolution in whole body layer images obtained with the detector that has full-width elements. Besides the obvious disadvantage of having to use two different detectors, an additional disadvantage is that a mechanism must be provided for moving one of the detectors out of the b am path and the oth r detector into it which complicates and

increases the cost of the apparatus. Moreover, mechanisms usually have free play and backlash which mak s it difficult to get both detectors properly aligned with the x-ray beam as is required for the computer to reconstruct head, other narrow parts and whole body layer images accurately.

One problem dealt with by the present invention is to enable obtaining high resolution x-ray images for narrow body parts such as heads and also for the whole body without requiring that the x-ray tube and detector be shifted nor that one detector be substituted for another.

To accomplish this, in accordance with one aspect of the invention, an x-ray detector is provided with a group of individual detector elements in its central region which have lesser width than detector elements in groups on each side of the central group. If, by way of example, the central group elements had one-half the width of the elements in the outside groups, there would be twice as many elements in the central group as there are in the two outside groups. For whole body views having a particular resolution, the full width elements in the outside groups are used and the narrower elements in the central group are connected in parallel so that if the element size ratio is 2 to 1, two elements in the central group provide an effective element width that is equal to one element in either of the symmetrically arranged outside groups. For taking views of head layers which only project onto the central group, the elements in the central group are used individually, that is, not in parallel in which case the total number of elements in the central group for a head view is equal to the total number of elements in the combined groups when views of whole body layers are being taken. Basically the detector elements are switched into proper arrangements for getting the desired resolution for head and narrow body parts and whole body views and for obtaining high resolution views of parts of the anatomy within a whole body layer while at the same time displaying the anatomy on each side of the high resolution part at lower resolution. Thus, all of the x-ray data acquisition channels can be used for heads and narrow anatomical regions as well as whole body layers and higher resolution images for heads can be obtained as compared with what would be obtainable if fewer of the wider detector elements were used for heads.

Another problem dealt with by the invention is to provide for obtaining and displaying a high resolution image of an anatomical region of the body which lies within a body layer. The region would have about the same size as a high resolution head image. The mode now under discussion is analogous to zooming in on a particular zone and displaying it at high resolution. In this mode, the narrow detector elements in the central group are used individually, that is, they are not connected in parallel and it is only the central group of elements which are used for the high resolution image within the image reconstruction circle. The elements on either side of the central group are used for obtaining data which the computer requires for proper reconstruction of the high resolution zone.

In addition to the basic detector element selection and switching schemes outlined above for displaying high resolution head or narrow body part images, normal resolution whole body layers and high resolution or zoom images lying within a whole body layer where the lower resolution images on each side of the high resolution image is optionally displayed or not displayed, the invention provides for switching detector elements and using some of the elements which are outside of the image reconstruction circle or zone to provide data to the computer relative to parts of the body or non-anatomical objects such as catheters, medical dressings, parts of casts and so forth, which lie outside of the image reconstruction zone but are in the x-ray beam at least some of the time during scanner rotation. The computer reconstructs and displays the part of the image which is defined in the computer algorithm. However, since the body or a part of it such as the head is not circular but is wider in one direction than in another, there will be times during a rotational scan of the x-ray source and detector when a narrow part of the body fits within the boundaries of the diverging beam spaced with space to spare on the detector array and other times when the image extends over the whole detector and beyond the reconstruction zone. Also, some non-anatomical objects may lie outside of the image reconstruction zone so some detector elements are reserved for providing signals to enable faithful reconstruction even though attenuating matter lies outside of the image reconstruction zone.

In solving these problems, in accordance with the present invention, the detector elements or the attenuation data from them can be switched such that when a high resolution head image, for example, is being obtained with the more densely packed central group of detector elements providing the essential attenuation data within the reconstruction zone, other elements on either side of the central group are used as auxiliary elements so total head size becomes known to the computer. As another example, when a whole body layer of a patient falling within one maximum size range is being scanned, the central group of less than full width elements will be effectively paralleled in pairs to provide normal resolution and some full width elements in second groups on each side of the central group will also be used for providing useful data within the reconstruction zone or circle

for this body siz range and other full size el ments outward from the second groups will be switched to serve as auxiliary elements.

In an embodiment of the invention described in detail, instead of using switching circuits to parallel the less than full width detector elements in the central group, all of the signals from the less than full width and full width elements are fed to a general purpose digital processor which, under the control of the host computer, manipulates the data as required to produce the effect of adjacent elements in the central group having been switched to form element multiples each having a total width equal to a full width element in a side group as in the case of whole body views, or manipulates the data to produce the effect of the elements in the central group being used individually as in the case of high resolution head views. In other words, the signals necessary for any of the modes and auxiliary element assignments are determined by the computer rather than by selective electronic switching of the data channels.

How the foregoing and other more specific objects of the invention are achieved will be evident in the more detailed description of illustrative embodiments of the invention which will now be set forth in reference to the drawings.

Figure 1 is a schematic diagram of a computer tomography system in which the new detector system may be incorporated;

Figure 2 is block diagram of an x-ray detector array for explaining how certain detector elements are activated and inactivated for head layer and body layer x-ray views;

Figure 3 is a fragmentary diagram for illustrating how the detector elements in the central group have half the width of adjacent detector elements in adjacent or outside groups;

Figure 4 is a block diagram of one system for using x-ray detector elements in paralleled pairs or individually;

Figure 5 is a block diagram of another implementation of the detector system wherein a computer effectuates paralleling and individualizing analog signals from detector elements depending on the mode in which computed tomography apparatus is operating;

Figure 6 is a diagram for explaining how the detector is used in performing the high resolution head layer imaging mode;

Figure 7 is a diagram for explaining how the detector is used in performing the normal resolution whole body layer imaging mode;

Figure 8 is a diagram for explaining how the detector is used in performing the zoom or close-up mode wherein a high resolution image of a part of a whole body layer is displayed and it is also used to explain how the detector is used in a mode wherein a limited region in a whole body layer is displayed at high resolution while adjac nt regions in the same layer are simultaneously displayed at low resolution;

Figure 9 is a diagram for explaining how the detector is used in performing the mode wherein a high resolution image of the major part of a large head layer that falls within the reconstruction circle is displayed and parts extending outside of the circle are not displayed; and

Figure 10 is a diagram for explaining how the detector is used in performing the mode wherein a whole body layer is so large overall that parts extend outside of the image reconstruction circle and only the major part within the circle is displayed.

Fig. 1 is a diagram which incorporates the main components of a typical computed tomography system. A body which is to be subjected to a scan with a transversely directed x-ray beam for the purpose of producing an axial view of a layer of the body is designated generally by the reference numeral 10. A circle 11 represents the head which is smaller than the part of the body below the shoulders. When making an x-ray examination, the body is recumbent on an x-ray transmissive table top, not shown, so as to maintain the body in a horizontal attitude coincident with a longitudinal axis which is perpendicular to the plane of the drawing in Fig. 1 and is marked with the numeral 12. The x-ray scanner comprises an x-ray tube 13 on one side of axis 12 and a multi-element x-ray detector 14 which is on the side opposite of axis 12 from x-ray tube 13. The x-ray tube 13 and detector 14 are mounted on a rotatable scanner frame, not shown, and the tube and detector are driven for orbiting jointly about body 10 and, in particular, in circular paths which are concentric with longitudinal axis 12. In some CT systems, the x-ray tube is energized and is emitting a beam of x-rays continuously during an orbital scan and in other CT systems the x-ray tube is pulsed on and off at uniform angular increments of rotation or orbital movement for the purpose of stopping motion when x-ray attenuation data is being collected for each view of a body layer.

Two x-ray beam collimator blades 15 and 16 are depicted in Fig. 1. These blades provide an aperture 17 for collimating the x-ray beam into a fan shape in this example. The focal spot on the x-ray tube target is marked 18. The boundary rays of the x-ray beam as defined by collimator blad s 15 and 16 are marked 19 and 20 and, as can be seen, they diverge from focal spot 18 so that the beam spreads substantially over the entire length of multi-element detector array 14. The collimator has another pair of blades, not shown, which limit the thickness of the fan-shaped x-ray beam to about 1 centim ter in the dir ction measured along longitudinal axis 12.

X-ray detector 14 is comprised of a plurality

of detector elements which are represented by a plurality of radial lines and are collectively designated by the reference numeral 25 in Fig. 1. When the system is energized and the x-ray beam between boundaries 19 and 20 is projected through the body, attenuation of the spatial distribution of ray bundles or photon intensities across the fan-shaped beam as they emerge from the body 10 is detected by respective detector elements 25 which produce analog electric signals corresponding with attenuation along the ray paths which the elements intercept. The analog signals corresponding with photon intensitites are customarily fed into a data acquisition system 26 which filters, preamplifies and digitizes the signals and sends them to a computer or central processor unit (CPU) 27 by way of a suitable bus 28. The attenuation data obtained for each x-ray view taken as the x-ray tube and detector orbit the body is used by CPU 27, controlled by a suitable algorithm, to determine the x-ray attenuation by all of the individual body elements and to produce picture element signals (pixels) corresponding with the attenuation. These digital signals, representative of picture elements, are fed by way of a bus 30 to a conventional display controller 29 where they are held in its internal memory matrix. The display controller 29 causes the matrix to be read out on a line-by-line basis and converts the individual pixel signals to a corresponding analog video waveform which is fed by way of a bus 31 to a raster scanned video display or television monitor 32 which displays on its screen 33 the reconstructed image 34 of the body layer which has been scanned. The image is depicted as being circular since only that x-ray attenuation data taken during a scan within a reconstruction circle is used in the display. The image is crosshatched to suggest that it is comprised of pixels 35 which are light or dark depending on the attenuation by the small body volume elements which they represent. The CPU 27 has various memory devices 36 associated with it. One memory device may be for holding the x-ray attenuation data as it is being accumulated in preparation for execution of the reconstruction algorithm. Another memory device may be a magnetic disk for storing the pixel data representative of one or more body layers so that it can be accessed and sent to the display controller and video display at any time.

Block 37 is representative of a known type of x-ray control for energizing the x-ray tube and for pulsing it on and off in a pulsed scanning system and for controlling th cur- r nt and voltage of the x-ray tube.

As shown in Fig. 1, when a layer of a narrow body part such as the head 11 is being scanned, the image or shadow of the head falling on multi-element x-ray detector 14 will fall only on the central group of detector elements 25 lying between the diverging boundary rays which are spanned by the arc designated H. The same would be true if an arm or a leg or an infant were being examined. In such cases, most of the detector elements in the groups on the opposite sides of the H region are not involved in providing analog signals relevant to image reconstruction of the head layer or of a less than full body width region in the anatomy. However, some of the detector elements to the right and left of the head or H group are used for other purposes. For instance, those elements which are immediately inside of boundary rays 19 and 20 are used to provide signals indicative of the constancy of the x-ray beam intensity which could fluctuate a little with line voltage fluctuations. The reference signals obtained from these detector elements provide a base line or signal level to which the other useful attenuation indicative analog signals may be referenced or normalized as is well-known. Other detector elements 25 immediately adjacent the rays H which bound the head may be used to produce signals which the CPU needs to determine the true size of the head and which provide isolation from the other detector elements.

When a view of the whole body (WB) is being taken, the boundary rays which just graze the body are at the limits of the arc marked WB. One may see that the image of the whole body layer is projected on a larger number of detector elements 25 than is the case when only the head, for example, is being scanned. During whole body layer scanning, useful data for image reconstruction is obtained from detector elements lying between opposite diverging rays 40 and 41. These rays are tangent to the image reconstruction circle which is not represented in the drawing. Elements lying between rays 41 and the next limiting ray WB are considered as auxiliary elements which produce signals that indicate the maximum size of the body to the computer or permit recognition of external non-anatomical objects such as medical dressings, casts and so forth by the computer. For instance, if the x-ray tube and detector were rotated 90° from their positions in Fig. 1, the body layer would look thinner because of the body being thinner in the horizontal direction. Moreover, the computer algorithm requires data such as for correcting for an arm not being in the x-ray beam when the tube and detector are in a particular angular position but being in the beam in the other angular positions. Reference signal producing detector elements at opposite ends of detector 14 are, however, also used when making a scan of a whole body lay r. The main purpose of the foregoing detailed discussion is to indicate the conventional practice wherein all of the detector elements 25 have the same width, such as two millimeters, and wherein only a central

group of elements is used for imaging narrow regions such as heads and a substantially greater number of elements in groups on opposite sides of the central group are used with the central group for taking views of whole body layers. In this prior art detector system wherein all detector elements which are used for either whole body layer images or head layer images have the same width, images of either type will necessarily have the same resolution.

The detector elements 25 may be of the gas ionization chamber variety or they may be solid state scintillation detectors, not shown, if desired. In any case, their function is simply to detect the photon intensity distribution across the beam and produce analog signals representative of attenuation along the various ray paths that comprise the x-ray beam and to collect a group of signals simultaneously for each rotational angular position of the jointly orbiting x-ray tube and detector element array. An exemplary gas ionization type detector is shown and described in detail in U.S. Pat. 4,161,655 which is owned by the assignee of this application and is incorporated herein by reference.

The new dual function x-ray detector for obtaining attenuation data for images of narrow body portions such as the head and whole body layers of equal resolution and for obtaining data for high resolution images within lower or normal resolution portions is shown schematically in Figs. 2 and 3 and data processing systems for cooperating with the detector are shown in block form in Figs. 4 and 5.

In Fig. 2, the new x-ray detector is designated generally by the reference numeral 44 and is comprised of a series of detector elements which are not shown individually in this Fig. although the boundaries of various groups of them are shown. It may be considered that the array of detector elements extends from one outside boundary or wall 45 to the other 46. The detector elements are shown as being classified in groups in Fig. 2. The groups are designated by the letters R, C, B, A, H, A', B', C', and R'. Another grouping consists of R, AUX, WB, AUX' and R'. The cenral group H is comprised of detector elements which have half the width and, hence, have twice the resolution capability of the elements in the other groups. This is illustrated in Fig. 3 where the elements, such as the one marked 47 bounded by separating lines 48 and 49 in the central group H are half as wide as the lements, such as the one marked 50, which is bounded by the lines 51 and 52 in adjacent group A'. This detector could be of the gas-filled ionization type or of the solid state type. For instance, elements 47 and 50 can be prisms of a material which scintillate and produce light in response to absorption of x-ray photons.. At the edge of each scintillation crystal ther is a light sensitive diode such as the typical ones marked 53 and 54 which produces a current on their output lines, such as the one marked 55, that is an analog of the x-ray photon intensity intercepted by a particular element.

The detector in Fig. 3 may also be looked upon as a gas-filled ionization type in which case the spaces such as 47 and 50 would be occupied by gas which produces electron-ion pairs in response to absorption of x-ray photons. In a gas-filled type, the elements 53 would constitute electrodes as in the patent cited earlier and they would have lead wires 55 for conducting the analog signals that correspond with the x-ray intensity. By way and not limitation, the widths of the detector elements in group H are typically one millimeter and the widths in the other groups are two millimeters so the elements in central group H are one-half the width of the elements in the other groups. Stated more generally, the elements in the outside groups must be wider than those in the central group.

In Fig. 2, the boundary rays of the fan-shaped x-ray beam coming in toward detector 44 are marked 19 and 20 as they are in Fig. 1. The boundary rays which pass through the sides of the head 11 define the useful image reconstruction zone or circle of the head and these rays and the rays between them are intercepted by the less than full width detector elements in central group H. In the head examination or high resolution mode, as will be explained in greater detail later, the less than full width or half-width detector elements, in this example, in group H are connected to effectively yield individual analog signals from the half-width detector elements. In the high resolution mode, the full width detector elements in groups A and A' are used as auxiliary detector elements to provide signals which indicate to the CPU the maximum width of the head, for instance. The signals from these elements assure that artifacts will not result in the reconstructed image as a result of a part of the body being in the x-ray beam during only part of a scan revolution. In the head mode, groups of elements B, C, B' and C' are inactive or not used. In either the head mode or whole body mode, however, at least one element in the x-ray beam intensity monitoring or reference groups R and R' are used and they provide for monitoring beam energy fluctuations, if any, and enable all signals to be normalized relative to the instantaneous reference level.

In the whole body mode, the half-width detector elements in central group H are connected or used as if pairs of them were connected in parallel to become equivalent to full-width elements and the full size elements at opposite sides of the central group, that is, the elements in groups A, B, A', and B' are used to produce the effect of having the entire

detector being made up of full-width elements. The effective elements for the whole body mode are treated as a unified group which is designated by the letters WB in Fig. 2. The groups C and C′ correspond with the groups AUX and AUX′ and constitute the auxiliary groups in the whole body mode as indicated and as will be discussed more fully later. A detector element or more in each of the reference groups R and R′ is also used in the whole body mode. Thus, for the whole body, all the detector elements in group WB are used for image reconstruction and they fall within the boundary rays 40 and 41 which are similarly identified in Fig. 1.

The easiest way to demonstrate how the detector elements are used for the head view and body view modes is to give illustrative numerical values to the number of detector elements in each of the groups. This is intended to be illustrative rather than limiting. Disregarding reference signal elements, consider that there are a total of 720 cells used for producing signals that contribute to the displayed image and that different combinations of elements are used for the various modes. The number of elements in each group are designated by the numbers which are in parentheses in Fig. 2. Thus, groups B and B′ contain 90 full width, 2mm in this example, detector elements. Groups A and A′ each contain 30 full width or 2mm elements. The central group H contains 480 half-width or 1mm elements which can be paralleled in effect, to produce 240 full-width elements. It will thus be evident that for the whole body, WB, 90 + 30 + 240 + 90 + 15, making a total of 480 elements in use. The elements in the auxiliary groups AUX and AUX′ can be disregarded for the moment because the analog signals from them only provide information which is used in the image reconstruction algorithm by the computer to determine the true and maximum size of the body for reasons which were given before.

Use of the central group H half-width elements as paralleled pairs so they act like full-width elements in the whole body mode is illustrated with the switching circuit 60 in Fig. 3. The one switching circuit which is shown has one switch 61 that is presently closed and another switch 62 that is open. Hence, the detector elements 63 and 64 to which this switching circuit connects are connected in parallel through switch 61 and effectively become a single detector element comprised of the two half-width elem nts 63 and 64. The analog signal, representative of x-ray photon intensity, is taken out by way of line 65 and further processed as will be explained. Referring further to Fig. 3, using half-width detector elements 63 and 64 individually as is done in the head mode, requires that switch 61 be opened and that switch 62 be closed so that the individual signals from half-width detector elements 63 and 64 can be outputted for further processing on individual output lines 65 and 66. Examples of how switching is actually done for the head and body modes and the composite high and low resolution mode will be discussed in greater detail later.

Referring again to Fig. 2, for the head mode, the 480 half-width detector elements in group H are used individually or unparalleled and constitute the image reconstruction circle or zone falling between limiting rays 47 and 48. It should now be evident that in the head mode the group H which constitutes the image reconstruction zone has 480 active detector elements. Similarly, for the whole body mode, 480 detector elements are in use by virtue of using the 30 elements in groups A and A′, the 90 elements in groups B and B′ and 240 unparalleled or full-width elements in group H. Hence, the spatial resolution for heads is two times higher than for bodies, that is, the picture element resolution obtainable in the whole body and head modes is the same.

Stated in more general terms, for the whole body mode, the central group H has a predetermined number of detector elements and each of groups A + B and A′ + B′ contain half as many elements as the central group H. Thus, when the number of detector elements in group H are double as is effectively done in this example in the head mode by using the elements individually, the number of effective elements in this mode is the same as the number of elements in the whole body mode. Stated more generally, the detector elements in the central group all have a certain width and the widths of the elements on each side of the central group are preferably and most conveniently made equal to said certain width multiplied by a whole number such as 2, 3 or 4.

Various imaging modes that be carried out with the system will now be discussed sequentially in reference to Figs. 6, 7, 8, 9 and 10 in conjunction, respectively, with Fig. 2.

The displayed head layer image 11′ in Fig. 6 results from operating in the high resolution head imaging mode. In this mode, the 480 half-width (or one millimeter wide elements in this numerical example) in the central group H of Fig. 2 are used individually to produce high or one millimeter resolution. The head size of this particular patient is shown as being small nough to fit with clearance within the field of vi w or image reconstruction circle which is shown surrounding the head layer image 11′ and is unmarked. In one design the image reconstruction circle has a diameter of 23cm as determined by the computer algorithm. When the head size is small enough to fit within the field of view or reconstruction circle it is not necessary to use any of the detector elements from the groups A, B on one side

and A′, B′ on the other side of central high resolution group H as auxiliary elements for informing the computer as to maximum head size since the whole head is in the reconstruction zone.

The displayed whole body layer image 10′ in Fig. 7 results from operating in the normal lower resolution mode. In this mode, 480 elements are used for the displayed image but they are effectively all 2mm elements. The total number is made up by, in effect, switching the 480 one millimeter individual elements in central group H into 240 parallel adjacent pairs of elements which then act as 240 individual two millimeter elements and by also using the 30 two millimeter elements in each of groups A and A′ and the 90 two millimeter elements in each of groups B and B′ which add up to 240 additional two millimeter elements and make up the total of 480 effective elements. As shown in Fig. 7, the entire whole body layer 10′ lies within the image reconstruction circle so it is not necessary to use any of the detector elements outside of groups B and B′ as auxiliary elements for determining maximum body size. By way of example, in one CT machine design, the image reconstruction circle or field of view is 46cm in diameter for the whole body layer mode.

Fig. 8 shows the result of effectively switching the detector elements in the zooming or close-up mode wherein a high resolution image of only a particular region within a body is displayed and attenuation data is taken from the whole body for determining body size but only the region for which high resolution is desired is displayed. Using the numerical example, 720 detector elements are in use in this mode and 480 are used for the displayed high resolution image. The 480 elements are the high resolution central group H of one millimeter elements. The difference between 720 and 480, that is, 240 are made up of the sum of the two millimeter low resolution elements in groups $A + B = 30 + 90 = 120$ on one side of group H and $A' + B' = 30 + 90 = 120$ on the other side of group H. Of course, the reference signal elements in groups R and R′ are in use as they are in all cases. In Fig. 8, the high resolution image is bounded by the image reconstruction circle 81. The shaded area 82 around circle 81 represents where the whole body layer would be if it were displayed. By way of example and not limitation, in said on CT machine design, the image reconstruction circle 81 or field of view in the body is 23cm in diameter and the resolution is, of course, 1mm.

Fig. 8 is also useful to illustrate another mode wherein a high resolution image 81 of a selected region in the body is displayed as in the preceding paragraph and, in addition, a lower resolution image of the contigous whole body regions represented by the shaded area 82 is also display d. The detector elements involved and the manner in which they are ffectively connected for this mode compares with the use of the elements in the preceding paragraph.

The foregoing discussion reveals how the detector elements are switched for producing high resolution head layer images or less than whole body or limited region, lower resolution whole body layer images and high resolution images of a particular body region in a whole body layer for normal size bodies or parts which fall entirely within the reconstruction circle. As will be explained later, in the preferred mode for processing the x-ray attenuation data derived from the detector elements, a computer is used to handle the data in a manner which simulates connecting the detector elements or switching the elements to enable display of images in any mode. The computer stores the digital value equivalents of the analog signals from the individual detector elements for each exposure and can manipulate the data to produce any selected type of image commanded by the operator. Thus, a meritorious feature of the system is that the x-ray exposures for as many layers as are required from the patient can be made and the patient can leave since the computer has the data stored. Consideration will now be given to the matter of switching detector elements which serve as auxiliary elements under various conditions such as for determining body or head size and which are used to cope with whole body layers and head layers which are large enough to have parts extend outside of the low and high resolution image reconstruction circles.

Refer to Fig. 9 which shows a large head layer image where the central group H elements are switched to the high resolution or individual 480 element condition and the auxiliary cells are switched for determining full head size, that is, the amount by which the head extends out of the high resolution imag reconstruction circle. By way of example, the head might be 30cm overall and the high resolution image reconstruction circle is, in this example, 23cm in diameter in Fig. 9 whereas in Fig. 6 it was assumed that the head was equal to 23cm or less so it would fall entirely within the predetermined 23cm reconstruction circle. In Fig. 9, the reconstruction circle is marked 83. Parts 84 of the head extend outsid of th circle and would be in the undisplayed area indicated by cross hatching and marked 85. In this case, a total of 720 detector elements are used in the following manner. The 480 one millimeter detector elements in group H are switched into the unpaired individual state to provide the attenuation signals for the high or one millimeter resolution image portion within circle 83. 240 one millimeter elements are nom-

inally switched and effectively paralleled in groups of 4 to become 60 four millimeter elements that serve as auxiliary elements. What happens is that the 30 elements in group A and 90 elements in group B are grouped in fours to make up 30 elements and similarly the 30 elements in group A' and 90 elements in B' are grouped to make 30 four millimeter elements effectively. Of course, the reference signal elements in groups R and R' ar used.

Fig. 10 illustrates the situation where the whole body layer 10'' size is greater than the image reconstruction circle for the normal resolution mode so that parts 86 of the body and possibly non-anatomical objcts lie outside the reconstruction circle and would not be displayed but are cross hatched to indicate where they would be theoretically. For example, the image reconstruction circle in the body would be 46cm in diameter as in the Fig. 7 example. The body for example, might have a total width of 60cm. 480 elements effectively are still needed to utilize all of the data channels for image reconstruction and for getting the best possible resolution. The number is achieved by effectively pairing the 480 one millimeter elements in group H to produce the equivalent of 240 two millimeter elements. The remaining 240 two millimeter elements required are made up by raising the 30 and 90 two millimeter elements in groups A and B for 120 and the 30 and 90 two millimeter elements in groups A' and B' for the other 120 elements. The 15 two millimeter elements in groups C and C', respectively, are then used as auxiliary elements for determining maximum body size.

Figs. 4 and 5 show two different systems in which the signals from the new dual detector can be used for imaging reconstruciton in the various modes.

In Fig. 4, the new multiple mode detector is designated generally by the reference numeral 44 as it is in Fig. 2. An incoming x-ray image is symbolized by the boundary rays 19 and 20 of the fan-shaped x-ray beam. The individual analog signals from the detector elements are conducted by way of a bus 70 to a low-pass filter and amplifier array 71 which filters and amplifies the analog signals in the chann ls or lines from the respective detector elements. These analog signals, of course, correspond with x-ray attenuation in the discrete ray paths within the thin x-ray beam for each x-ray view. A set of signals is produced for each x-ray view that is, for each increment of scanner rotation. In this arrangement, there is no switching of the detector elements in the central group H for the head or body modes until the signals have been amplified. This results in a better signal-to-noise ratio when the signals are processed further. The signals from the filter and amplifier array 71 are conducted by way of a bus 72 to an analog demultiplexer 73. The demultiplexer is a conventional type which performs the detector element switching function in this embodiment. That is, it is controllable to switch or route the analog signals in various ways for the head mode, whole body mode and zoom mode. In other words, the demultiplexer selects those detector elements which are to be used for image reconstruction and for auxiliary signal elements for the head and body layer modes as required. This is represented symbolically in Fig. 4 by showing that the analog signals involved in the body layer mode are outputted on a bus 74 and those which are used for the head mode are outputted on a bus 75 in Fig. 4. These buses join in a common bus 76 which is the input to a sequence multiplexer 77 which selects the particular channel or line out of the demultiplexer which is to be converted to a digital value next. The signals are fed in sequence by way of bus 78 to a group of analog-to-digital (A/D) converters which are symbolized by the block marked 79. Of course, there is a set of analog signals coming into the A/D converter 79 for each x-ray view or incremental change in the angular position of the x-ray tube and detector during a rotational scan of the body. The host CPU 27 reads out these sets of analog signals from the A/D converter via bus 80 synchronously with angular rotation of the scanner. The CPU uses the digital values in the image reconstruction algorithm and produces a set of digital signals which are representative of the intensity or brightness of the picture elements in the image which is to be displayed on the video monitor 32. In Fig. 4, the pixel representative digital signals are coupled by way of a bus 30 as in Fig. 1 to the display system, generally designated by one block which would be comprised of the display controller 29 and video display unit 32 as depicted in Fig. 1. In the Fig. 4 system, the analog signals are outputted from the demultiplexer 73 in a format which depending on the mode that is in effect, enables the CPU to determine which of the corresponding digital signals are image producing signals, which are auxiliary element signals and which are reference level signals.

In the Fig. 4 system, the output lines from the x-ray detector elements are actually switched or demultiplexed in accordance with whether the whole body layer mode, zoom mode or head layer mode is in effect. In the preferred Fig. 5 system, on the other hand, no actual switching is done up the line but the computer is programmed so that is becomes an instrumentality which selects signals in accordance with the selected mode and determines if those derived from the central element group H should be combined in pairs to produce the effect of a single signal from a pair of elements as in the whole body mode or whether the signals from the half-width

elements should be treated individually, that is, not added or effectively paralleled as is the case in the zoom mode and head mode. The Fig. 5 system allows obtaining any combination of detector elements which might be desired for various applications.

The Fig. 5 preferred data processing system includes the x-ray detector 44 for supplying analog signals from the detector elements by way of a bus 90 to filter and amplifier array 91. In Fig. 5, however, the amplified analog signals from the detector elements are fed directly to A/D converter 93 where they are converted to 16 or 32-bit digital bytes, for example. These bytes are transferred by way of bus 94 to the internal memory of a digital processor which is designated generally by the reference numeral 96. The processor may be a general purpose type or an array processor. The particular calculations and other operations carried out by the processor is governed by the host CPU 27 and the control signals it sends to the processor by way of control bus 98 and this depends on which of the imaging modes is in effect. The processor does such things as pairing or not pairing the x-ray attenuation data signals from the central group of detector elements depending on whether normal resolution or high resolution image modes are in effect. In other words, instead of switching data signals to different channels up the line as in the Fig. 4 system, the processor in Fig. 5 has a full set of signals available from each x-ray view and it produces the effect of switching by properly pairing or not pairing signals and by selecting those signals from detector elements which are to be treated as auxiliary element signals. As the signals for each x-ray view are processed they are transferred by way of bus 99 to a disk memory 100 where they become available to the CPU 27 by way of bus 101.

The CPU uses the data from the disk memory to reconstruct the image in the mode desired and format its data for transfer to the display system comprised of display controller 23 and video monitor 32.

Although the new multiple function detector and associated signal processing systems have been described in considerable detail, and although specific numbers and various combinations of detector elements have been indicated for making explanation easier, such description is intended to be illustrative rather than limiting, for the detector and signal processing systems may be variously embodied without departing from the spirit and scope of the invention.

CLAIMS

1. A computed axial tomography system including x-ray source means for projecting an x-ray beam through a body layer and x-ray detector means comprised of a plurality of adjacent detector elements for detecting radiation emerging from the body, said elements being operative, resp ctively, to effect production of analog signals representative of x-ray attenuation in substantially discrete ray paths across the beam for various angles of the beam relative to the body, and means for processing said signals to enable displaying a reconstructed image of the layer characterized by an

improved detector means for said system for enabling obtaining and displaying high resolution images of relatively small anatomical regions such as legs, infant torso and head layers or a small region within a comparatively larger anatomical region and relatively lower resolution displayed images of said larger anatomical regions, said detector means including:

a plurality of adjacent detector elements of one width composing a central group and a plurality of adjacent detector elements of greater width than the elements in the central group arranged on each side of said central group, and

means for coupling signals produced by said detector elements to said means for processing.

2. The apparatus as in claim 1 wherein the elements in said central group are one-half as wide as the greater width elements on each side of said central group.

3. The apparatus as in claim 1 wherein said means for coupling said signals to said means for processing includes means operative to effectively connect predetermined numbers of adjacent elements in said central group in parallel to produce effectively larger central group element widths.

4. The apparatus as in claim 1 wherein said means for coupling said signals to said means for processing includes means operative to effectively connect adjacent pairs of the individual elements in said central group in parallel to produce central group elements which have effectively twice the width of the individual elements in the central group.

5. The apparatus as in claim 1 wherein said means for coupling said signals to said processing means includes means operative to connect effectively adjacent pairs of elements in said central group in parallel to produce effective central group element widths equivalent to the widths of the respective elements on each side of the central group.

6. The apparatus as in claim 2 wherein the group of elements on each side of said central group each contains equal numbers of greater width elements and the sum of the elements in said side groups equals the number of pairs of half-width elements in said central group.

7. In a computed axial tomography system according to Claim 1 including x-ray source m ans for projecting an x-ray beam through a layer of a body and x-ray detector

means comprising an array of adjacent detector elements for detecting radiation emerging from said body where radiation passing through substantially the whole body width is intercepted by a greater extent of the array than radiation passing through a layer in a region in the body which has substantially less than whole body width, said elements responding to radiation by respectively producing analog signals representative of x-ray attenuation variations across the emerging beam for a plurality of angles of the beam relative to the body, means for processing said signals for reconstructing the image and for producing signals representative of the picture elements which compose an axial image of said layer, and display means responsive to said picture element signals by displaying said image of said layer;

the improvement which enables obtaining a high resolution image in the case of a layer such as in the head or in a leg or arm which has less than whole body width and a relatively lower resolution image of a layer which has substantially whole body width and results from said radiation being projected on a greater extent of said detector element array than for said case wherein:

said detector element array is comprised of a central group of adjacent individual relatively small detector elements and groups comprised of adjacent detector elements on opposite sides of said central group, the widths of the elements in the central group being less than the widths of the relatively larger elements in the groups at the sides of said central group, and

means operative when a high resolution image of a layer of less than whole body width is desired to select the signals from the relatively smaller elements in said central group individually for reconstructing an image of said layer.

8. The system as in claim 7 wherein when an image is desired for a whole body layer whose projected radiation extends over a greater number of detector elements than are in the central group, said means are operative to select simultaneously the signals from the r latively larger elements in the side groups and from adjacent pairs of the relatively smaller elements in the central group which thereby act as larger width elements in the central group for reconstructing an image of said layer.

9. The system as in claim 7 wherein when a high resolution image of a region within a whole body layer is desired where the projected radiation ext nds over the central group of elements and said side groups and said region is within the limits of said central group, said means is operative to select the signals from the individual elements, respectively, in the central group for reconstruction and display of said high resolution image and said means is operative to s lect the signals from the respective relatively larger elements in said outside group for determining the size of said layer.

10. The system as in any of claims 7, 8 or 9 wherein said means for processing said signals includes:

filter and amplifier means having input means for the respective analog signals from said detector elements and output means for said signals after they have been filtered and amplified,

analog signal demultiplexer means having input means for said respective signals and having output means and being operative to selectively switch the signals from the individual detector elements in the central group and signals from said relatively larger detector elements in said side groups into said output means,

a sequence multiplexer means having input means for the signals from said demultiplexer means and having output means, said sequence multiplier being operative to select signals on its input means and couple them to its output means,

analog-to-digital converter means having input means for the selected analog signals and having output means for corresponding digital signals, and

means for using said signals for producing signals representative of a reconstructed image of a layer of the body or a part thereof.

11. The system as in any of claims 7, 8 or 9 wherein said means for processing said signals includes:

filter and amplifier means having input means for respective analog signals from said detector elements and output means for said signals after they have been filtered and amplified,

analog-to-digital converter means having input means coupled to the aforesaid output means for receiving said analog signals and having output means for corresponding digital signals,

digital processor means having input means for receiving the respective digital signals and having output means, said processor means being operative to select signals from said elements in said central group and from said relatively larger elements in accordance with the type of high resolution or low resolution image that is to be reconstructed, and

means for controlling said processor means to make the selection.

12. A computed axial tomography method for obtaining x-ray attenuation data that enables reconstruction of the highest resolution images of layers within whole bodies, regions within whole bodies and parts which have less than whole body width that an x-ray detector of predetermined length will permit in a tomography system wherein an x-ray beam from an x-ray source is projected through a

layer of the body and the attenuated radiation which emerges from the body layer is detected by a detector that has an array of adjacent detector elements which produce analog signals corresponding with the intensity of the radiation they detect and wherein said detector and source are jointly rotated around the body to obtain attenuation data at a plurality of rotational angles, said method involving:

having the detector elements arranged as a central group of narrow equal width elements with a series of relatively wider elements on each side of said central group,

selecting signals from the individual detector elements in the central group for reconstructing a high resolution image of a region of a body layer whose emergent radiation extends substantially over the length of the central group, or alternatively

selecting signals resulting from combining signals from pluralities of adjacent narrow elements in the central group and signals from a plurality of wider elements on each side of said group for reconstructing a relatively lower resolution image of a wider boy layer whose emergent radiation extends over a length of the detector greater than the length of the central group.

13. The method as in claim 12 wherein when signals for a high resolution image are being selected from said central group of narrow elements some of the wider elements on each side of said group are used as auxiliary elements whose signals are used for eliminating the effects of external non-anatomical objects and for determining the overall width of the body layer.

14. The method as in claim 12 wherein when signals for said lower resolution image are being selected, wider elements on each side of the wider elements whose signals are used for the image are used as auxiliary elements whose signals are used for eliminating the effects of external non-anatomical objects and for determining the overall width of the body layer.

15. A computed axial tomography method for obtaining x-ray attenuation data that enables reconstruction of the highest resolution images of layers within whole bodies, regions within whole bodies and parts which have less than whole body width and that an x-ray detector of predetermined length will permit in a tomography system wherein an x-ray beam from an x-ray source is projected through a layer of the body and the attenuated radiation which emerges from the body layer is detected by a detector that has an array of adjacent detector elements which produce analog signals corresponding with the intensity of the radiation they detect and wherein said detector and source are jointly rotated around the body to obtain attenuation data at a plurality of rotational angles, said method involving:

having the detector elements arranged as a central group of narrow equal width elements with a series of relatively wider elements on each side of said central group,

selecting signals from the narrow individual detector elements in the central group for reconstructing a high resolution image of a limited region of a body layer whose emergent radiation extends substantially over the length of the central group and simultaneously selecting signals from the wider elements in the side groups for coincidentally reconstructing a lower resolution image of the parts of the body outside of said region, or alternately,

selecting signals resulting from combining signals from pluralities of adjacent narrow detector elements in the central group for reconstructing a low resolution image of a limited region of a body layer whose emergent radiation extends substantially over the length of the central group and simultaneously selecting signals from the respective wider elements on the sides of said central group for coincidentally reconstructing a correspondingly low resolution image of parts of the body whose emergent radiation is projected on the wider elements on each side of the central group.

16. A computed axial tomography system as claimed in claim 1 including an x-ray detector substantially as described herein with reference to Fig. 3 of the accompanying drawings.

17. A computed axial tomography system as claimed in any one of the preceding claims including a signal handling arrangement substantially as described with reference to Fig. 4 or Fig. 5 of the accompanying drawings.

18. A computed axial tomography syst m including features disclosed herein in any novel combination.

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